"Express Mail" Mailing Label Number EL852677641US
Date of Deposit: JUNE 20, 2001
I hereby certify that this paper or fee is being deposited with the United States Postal Service "Express Mail Post Office to Addressee" service under 37 C.F.R. 1.10 on the date indicated above and is addressed to: Assistant Commissioner for Patents Washington, D.C. 20231.
By: Hilary M. McNULTY

MULTIPLE CONTRAST ECHO-PLANAR IMAGING FOR CONTRAST-ENHANCED IMAGING

This application claims the benefit of U.S. Provisional Application No. 60/285,543, filed on April 20, 2001.

Background of the Invention

The present invention relates to the magnetic resonance arts. It finds particular application in medical diagnostic imaging and will be described with reference thereto. However, it will be appreciated that the invention will also find application in other types of imaging, spectroscopy, and the like.

A typical magnetic resonance (MR) imaging sequence includes an RF excitation pulse, e.g. a 90° pulse, with a corresponding slice- or slab-selective magnetic gradient pulse, followed by a series of spatial encoding and readout magnetic field gradient pulses. In some sequences a second 180° refocusing pulse is applied between the initial excitation pulse and the spatial encoding/readout region. The 180° pulse effectively reverses the dephasing effect of small spatial variations in the MR frequency due to spatial variations in the applied magnetic field, and refocuses the magnetization to form a spin-echo. MR imaging is performed using various imaging modes which usually vary with respect to the method and timing of the spatial encoding and data readout sequences.

The choice of spatial encoding and data readout scheme has significant consequences on the imaging contrast, resolution, and scanning speed. Two imaging parameters are the time-to-echo, T_E, and the repeat time between RF excitations, T_R. Sampling the induced resonance nearer to the excitation emphasizes proton density weighting or T_1 weighting in which the contrast strongly reflects the regrowth rate of the M_Z component of the net magnetization. Sampling the magnetization later emphasizes T_2 weighting in which the contrast strongly reflects the decay rate of the M_{XY} component of the net magnetization.

Proton density (ρ) weighting is obtained when the T_E delay is short and the magnetic resonance has minimal time to decay, so that the density of resonant hydrogen protons is measured. T_2^* weighting is obtained using a longer T_E delay so that the fastest (T_2^*) magnetic resonance decay is a factor. The T_2^* decay differs from T_2 in that T_2^* includes inhomogeneous dephasing due to static magnetic field inhomogeneities. To measure the "pure" T_2 corresponding to dephasing due to molecular interactions (excluding inhomogeneous dephasing), a 180° RF refocusing pulse is applied to induce a spin-echo during the sampling interval. Other types of pre-pulses can also be applied to provide fat suppression, MTC, et cetera.

Prior spatial encoding and readout schemes have been configured to provide a variety of ρ, T_2, or T_2^* weightings. The choice of spatial encoding scheme strongly affects the scan speed and resolution. A popular MR imaging mode is echo-planar imaging (EPI). In the EPI imaging mode, an oscillating read gradient generates a series of gradient echoes. Phase encoding pulses between echoes step the sampling through k-space in a back-and-forth rastering fashion. The speed of EPI is preferably sufficient that the k-space data for an entire planar (slice) image is obtained from a single RF magnetic resonance excitation, i.e. "single-shot" EPI, or SS-EPI.

The rapidly switched gradients along with a rastered readout timing sequence of SS-EPI produce complete slice scans in as little as a few hundred milliseconds or less. This speed makes SS-EPI an ideal method for clinical imaging when short scan times are important. Reduced scan times translate to reduced image blurring due to patient movements, respiration, cardiac action, and the like.

The EPI technique encompasses a number of variants, including several techniques collectively known to the art as partial parallel imaging (PPI). In the PPI techniques, a phased array receive coil simultaneously measures the MR response using a plurality of phased receive coils and combines the data from the array to acquire a plurality of k-space samples in parallel.

Enhancement in MR imaging can also be obtained through the use of multiple image techniques. In these methods, the spatial encoding scheme is designed so that multiple images, typically using more than one image contrast mode, are obtained from the echo train following a single RF excitation pulse. For example, a T_2^* weighted image and a T_2 weighted image can be obtained.

Another type of MR imaging is contrast-enhanced imaging. In this type of MR imaging, a magnetic contrast agent, such as a gadolinium chelate, is administered to the patient, such as by a bolus injection. The magnetic contrast agent provides enhanced MR contrast versus intrinsic imaging. In some studies, the preferential concentrating of the contrast agent in particular organs or tissues is imaged. In vascular imaging, the distribution of an administered contrast agent is monitored over time to study the performance of major blood vessels. Similarly, the perfusion of the contrast agent through tissues or organs enables study of the capillary performance in the targeted areas.

In order to quantitatively analyze perfusion by contrast-enhanced MR imaging, it is useful to quantify the concentration of the contrast agent in the imaged area

based upon the MR image. In the exemplary case, the gadolinium chelate strongly reduces the T_2 weighted signal and T_2^* weighted signal. In a T_2 weighted MR image, the areas of high gadolinium chelate concentration appear darker than the surrounding areas. In principle, therefore, the contrast agent concentration can be extracted from the percentage darkening or from similar quantitative image analysis. Unfortunately, competing effects, such as brightening due to T_1 shortening, can counteract the T_2 darkening effect of the gadolinium chelate and produce errors in the quantitative analysis.

The prior art also discloses taking a reference image prior to administration of the contrast agent. This approach has the disadvantage that the image of the contrast agent usually needs to be registered spatially with the reference image to correct for patient movement or other spatial shifting.

An effective method is needed for correcting these errors in quantitative contrast-enhanced perfusion imaging. Such correction would preferably utilize additional non-T_2 weighted images to account for extraneous, non-T_2 contrast mechanisms. However, the collection of these additional images is limited by the time constraints imposed by the dynamic perfusion process. The present invention contemplates a new imaging method which overcomes these limitations and others.

Summary of the Invention

According to one aspect of the invention, a method of magnetic resonance imaging is disclosed. A magnetic resonance contrast agent is administered to a subject, which contrast agent alters T_2 and T_2^* magnetic resonance characteristics. A magnetic resonance is excited in a region of interest of the subject which receives the contrast agent. A first echo planar image readout waveform is applied which generates first image data. After the first echo planar image readout waveform,

a second echo planar image readout waveform is applied and a T_2 or T_2^* weighted image data is generated. The image data is reconstructed to generate a proton density weighted or a T_1 weighted image representation and a T_2 or T_2^* weighted image representation. The T_2 or T_2^* weighted image representation is corrected with the first image representation.

Preferably, the method includes applying an RF inversion pulse between the first and second echo planar image readout waveforms.

The method preferably includes applying a third echo planar image readout waveform and generating the other of T_2 and T_2^* weighted image data. Optionally, an RF inversion pulse is applied between the second and third echo planar image readout waveforms, such that the second echo planar image readout waveform generates T_2^* weighted data and the third image readout waveform generates T_2 weighted data. The T_2 weighted data is preferably reconstructed into a T_2 weighted image representation, and the T_2 weighted image representation is preferably modified with the first image representation.

The method preferably includes reconstructing the T_2 or T_2^* weighted image data and a portion of the first image data to generate the T_2 or T_2^* weighted image representation, and reconstructing a portion of the T_2 or T_2^* weighted image data and the first image data to generate the first image representation. Optionally, the portion of the T_2 or T_2^* weighted data used in generating the first image and the portion of the first image data used in generating the T_2 or T_2^* weighted image include interleaved data lines adjacent an edge of k-space. Optionally, additional data lines are generated by conjugate symmetry.

Preferably, the method includes repeating steps of the method a plurality of times to generate a series of first image representations and a series of T_2 or T_2^* weighted image representations. These image series are

preferably combined to generate a third series depicting a temporal evolution of the contrast agent in the region of interest.

Preferably, the method includes combining the first image representation and the T_2 or T_2^* weighted image representation to generate a third image representation, and then repeating steps of the method a plurality of times to generate a series of third image representations depicting a temporal evolution of the contrast agent in the region of interest.

In the method, the contrast agent is preferably a gadolinium chelate.

At least one of the steps of generating the first image data and generating the second image data optionally advantageously includes generating image data using a partial parallel imaging technique.

According to another aspect of the invention, a method of contrast enhanced magnetic resonance imaging is disclosed. A subject is injected with a contrast agent, magnetic resonance is excited in a region of interest, the excited magnetic resonance is permitted to decay for a preselected duration to optimize one of T_2 and T_2^* weighting, and after the preselected duration an echo planar image readout waveform is applied to generate T_2 or T_2^* weighted data. The method further includes, during the preselected duration, applying another echo planar image readout waveform to generate T_1 weighted data.

According to yet another aspect of the invention, a method is disclosed for imaging a patient using a magnetic resonance (MR) imaging apparatus. The MR apparatus includes a patient support means, a main magnet, a slice-select gradient pulse generator, a phase-encode gradient pulse generator, a read gradient pulse generator, a plurality of RF coils, an RF transmitter, and a receiver. The method includes the steps of: administering a contrast agent to the patient; exciting a magnetic resonance in the patient using the RF transmitter

and at least one of the plurality of RF coils in conjunction with the slice-select gradient generator; encoding and reading the magnetic resonance using the phase encode and the read gradient generators in conjunction with at least one of the plurality of RF coils and the receiver, the encoding and reading implementing a first echo-planar image readout waveform; encoding and reading the magnetic resonance using the phase encode and the read gradient generators in conjunction with at least one of the plurality of RF coils and the receiver, the encoding and reading implementing a second echo-planar image readout waveform; and reconstructing the read, encoded, magnetic resonance into first and second image representations.

Preferably, the method further includes comparing the first image representation with the second image representation to obtain a third image representation thereby. Optionally, the method includes repeating the steps of exciting a magnetic resonance, encoding, reading, and reconstructing first and second images, and comparing the first images with the second images to obtain third images thereby. A temporal evolution of at least one of the first, second, and third images is determined.

In the step of reconstructing the second image, a portion of the phase and frequency encoded resonance from the first echo planar image readout waveform is preferably reconstructed into the second image.

In the method, the first echo planar image sequence phase encoding preferably includes phase encoding a first portion of the resonance such that a k_y component single-steps in a first direction, and phase encoding a second portion of the resonance such that the k_y component double-steps in the first direction. The second echo planar image readout waveform phase encoding preferably includes phase encoding a first portion of the resonance such that the k_y component double-steps opposite to the

first direction, and phase encoding a second portion of the resonance such that the k_y component single-steps opposite to the first direction. The reconstructing step preferably includes reconstructing the first and second portions of the first echo planar sequence and the first portion of the second echo planar sequence into the first image representation, and reconstructing the second portion of the first echo planar sequence and the first and second portions of the second echo planar sequence into the second image representation.

According to still yet another aspect of the invention, a magnetic resonance imaging apparatus is disclosed. A main magnet generates a temporally constant magnetic field through an examination region. An RF system excites and manipulates magnetic resonance in the examination region and receives and demodulates magnetic resonance signals from the examination region into data lines. A sorter is provided for sorting the data lines between a first data memory and a second data memory. A gradient magnetic field system generates magnetic field gradients across the examination region to spatially encode the resonance signals. A sequence controller: (i) controls the RF system to induce resonance; (ii) controls the RF and gradient systems to implement a first echo planar readout waveform which generates T_1 weighted data lines; (iii) controls the RF and gradient systems to implement a second echo planar readout waveform which generates one of T_2 and T_2^* weighted data lines, and (iv) controls the sorter to sort the T_1 and T_2 or T_2^* weighted data lines between the first and second data memories. A reconstruction processor reconstructs data lines from the first data memory into a first image representation and data lines form the second data memory into a second image representation.

The magnetic resonance apparatus preferably further includes a means for injecting a contrast agent into a subject in the examination region, and an image

processor for combining the first and second image representations into a contrast agent enhanced image representation. Optionally, the sequence controller controls the sorter to sort all of the T_1 weighted data lines and a portion of the T_2 or T_2^* weighted data lines into the first image memory, and all of the T_2 or T_2^* weighted data lines and a portion of the T_1 weighted data lines into the second image memory.

Preferably, the RF system further includes a phased array receive coil, and a partial parallel imaging (PPI) integrator which processes the readout of the phased array receive coil to generate data lines. The PPI integrator preferably processes the readout of the phased array receive coil using one of a simultaneous acquisition of spatial harmonics (SMASH) technique, a sensitivity encoding (SENSE) technique, and a parallel imaging with localized sensitivities (PILS) technique.

One advantage of the present invention is that it facilitates correction of extraneous MR effects.

Another advantage of the present invention is that it facilitates faster scan times.

Another advantage of the present invention is that it provides multiple images with complementary dynamic range.

Another advantage of the present invention is that it facilitates quantitative contrast-enhanced imaging.

Another advantage of the present invention is that it corrects for extraneous image contrast in perfusion imaging.

Yet another advantage of the present invention is that it separates out counteracting MR effects in contrast enhanced T_2 weighted SS-EPI imaging.

Still yet another advantage of the present invention is that it provides additional MR data for a given imaging time that can be processed or combined to

obtain improved and/or additional diagnostic information versus the prior art.

Still further advantages and benefits of the present invention will become apparent to those of ordinary skill in the art upon reading the following detailed description of the preferred embodiment.

Brief Description of the Drawings

The invention may take form in various components and arrangements of components, and in various steps and arrangements of steps. The drawings are only for the purpose of illustrating preferred embodiments and are not to be construed as limiting the invention.

FIGURE 1 is a diagrammatic drawing of a magnetic resonance imaging scanner for imaging fluid flow in accordance with one embodiment of the invention;

FIGURE 2 is a representative qualitative timing diagram for a single-shot echo-planar imaging sequence according to the prior art, which records a T_2 weighted image during a single repetition period T_R;

FIGURE 3 is a representative qualitative timing diagram for multiple contrast single-shot echo-planar imaging sequence according to one aspect of the invention, wherein a T_2 weighted image and a second image, which can be ρ weighted or T_2^* weighted, are recorded during a single repetition period T_R;

FIGURE 4 is a representative qualitative timing diagram for a multiple contrast single-shot echo-planar imaging sequence according to another aspect of the invention, wherein three images with ρ, T_2^* and T_2 weighting are recorded during a single repetition period T_R;

FIGURE 5 is a representative qualitative timing diagram for a multiple contrast echo-planar imaging sequence according to yet another aspect of the invention, in which data usable for the generation of two T_2 weighted images are collected during each single repetition period T_R;

FIGURE 6 is a representative qualitative timing diagram for a multiple field echo imaging sequence according to still yet another aspect of the invention; and

FIGURE 7 is a representative diagrammatic representation of the traversal of k-space with EPI data showing data sharing between images near an extremity of k-space during a multiple-contrast SS-EPI slice measurement.

Detailed Description of the Preferred Embodiments

With reference to FIGURE 1, a magnetic resonance imaging (MRI) scanner **A** includes a main magnetic field control **10** that controls superconducting or resistive magnets **12** such that a substantially uniform, temporally constant main magnetic field B_0 is created along a z-axis through an examination region **14**. An imaging experiment is conducted by executing a magnetic resonance sequence with the subject being imaged or examined (e.g., patient, phantom, or otherwise) placed at least partially within the examination region **14**. The magnetic resonance (MR) sequence includes a series of RF and magnetic field gradient pulses that are applied to the subject to invert or excite magnetic spins, induce magnetic resonance, refocus magnetic resonance, manipulate magnetic resonance, spatially and otherwise encode the magnetic resonance, to saturate spins, and the like.

More specifically, gradient pulse amplifiers **20** apply current pulses to a whole body gradient coil assembly **22** to create magnetic field gradients along x, y and z-axes of the examination region **14**. Typically, these include a slice-select magnetic field gradient generator acting in the z-direction to produce the slice-selective (G_z) gradient. A phase-select magnetic field gradient generator typically acts in the y-direction to produce the phase-selective (G_y) spatial encoding gradient. A frequency-select magnetic field read gradient generator

typically acts in the x-direction to produce the frequency-selective (G_x) spatial encoding gradient. Of course, the designation of the gradients is preferably user-selectable (e.g., for a particular experiment the y-gradient can be designated as the frequency-selective gradient while the x-gradient can be designated as the phase-selective gradient). Additionally, gradient generators can typically be combined to produce gradients along directions other than the three principle (x, y, z) directions, so that, for example, the slice-selective gradient can be generated along a direction at an acute angle to the z-axis through a combination of generators.

An RF transmitter **24**, preferably digital, applies RF pulses or pulse packets to a whole-body RF coil **26** to transmit RF pulses into the examination region. A typical RF pulse is composed of a packet of immediately contiguous pulse segments of short duration which taken together with each other and any applied gradients achieve a selected magnetic resonance manipulation. The RF pulses are used to saturate, excite resonance, invert magnetization, refocus resonance, or manipulate resonance in selected portions of the examination region.

For whole-body applications, the resulting resonance signals, generated as a result of a selected manipulation, are also typically picked up by the whole-body RF coil **26**. For imaging limited regions of the subject, local RF coils are commonly placed contiguous to the selected region. For example, as is known in the art, an insertable head coil (not shown) is inserted surrounding a selected brain region at the isocenter of the bore; a selected surface coil (not shown) is employed; or, other such specialized RF coils are employed. In addition to receiving RF pulses, optionally, the local RF coil can also transmit magnetic resonance signals into the selected region.

A phased array receive coil (not shown) can also be used for reading the resonance signals. As is known to

those of ordinary skill in the art, a phased array receive coil can be used in conjunction with one of the known partial parallel imaging (PPI) techniques to perform parallel k-space sampling. In these techniques, readout data is obtained in parallel by the component coils of the spatially distributed array, and the spatial information extracted from the combined phased coil array data partially replaces the spatial information ordinarily obtained using gradients. A number of PPI imaging techniques have been developed, including simultaneous acquisition of spatial harmonics (SMASH), sensitivity encoding (SENSE), parallel imaging with localized sensitivities (PILS), et cetera, and these techniques need not be described in detail herein for an enabling disclosure of the invention.

The resultant RF magnetic resonance signals picked up by one or another of the RF coils, or by the phased array receive coil, are employed and demodulated by a receiver **30**, preferably digital, to generate a series of data lines. Preferably, a sequence control circuit **40** controls the gradient pulse amplifiers **20** and the RF transmitter **24** to produce an MRI pulse sequence that generates magnetic resonance (MR) signals or echoes received and sampled by the receiver **30**. When using partial parallel imaging techniques, a PPI integrator **32** combines the signals from the individual coil components of the phased array (not shown) to obtain k-space samples in parallel according to known PPI techniques, e.g., SMASH, SENSE, PILS, et cetera, that account for the localized coil sensitivities and the spatial location of the component coils of the array.

In one aspect of the invention, the MRI scanner runs a single shot echo planar imaging (SS-EPI) sequence. SS-EPI is a rapid MRI technique which can be used to produce tomographic images, e.g., at video rates, and is particularly useful in perfusion and/or diffusion studies, for functional magnetic resonance imaging (fMRI),

et cetera. Preferably, the technique records an entire image in a single T_R period, where T_R represents the repeat time for the SS-EPI sequence.

A patient **42** receives a dose of a magnetic contrast agent **44**, such as a gadolinium chelate administered as a bolus injection. This contrast agent-assisted MR imaging is referred to as contrast enhanced imaging. The contrast agent **44** is preferably injected into the bloodstream which transports it through the body of the patient to a region of interest. The contrast agent **44** has distinctive magnetic resonance properties which enable identifiable imaging of the contrast agent **44** as it moves through the patient **42**. In the exemplary case the gadolinium chelate reduces the T_2 or T_2^* contrast (i.e., darkens the image) in areas where it has penetrated. However, as noted previously, competing factors such as T_1 shortening can produce an opposing brightening effect which complicates quantitative analysis of the movement of the contrast agent **44**.

In an exemplary imaging experiment, the gadolinium chelate **44** is transported into a region of interest in the brain of the patient **42**. The gadolinium chelate is preferably administered to the patient as a bolus injection. The spatial distribution of the concentration of the contrast agent in the brain region as it distributes through the processes of blood flow and perfusion into surrounding tissues is monitored during MR imaging. In this manner, the blood flow through the major blood vessels as well as the perfusion of the blood into the surrounding tissues is studied.

With continuing reference to FIGURE 1, the data lines collected by the receiver **30** and optionally processed by the PPI integrator **32** are sorted **50** in accordance with ρ, T_2, or T_2^* weighting and loaded into corresponding k-space memories 52_1, 52_2, 52_3. For greater speed, data lines are often collected in just over half of k-space. A conjugate symmetry processor **54** calculates

conjugately symmetric data lines to complete k-space. A reconstruction processor **56** reconstructs the k-space data into image representations, such as with a two-dimensional inverse Fourier transform or other reconstruction method as is known to the art. The resultant image representations are stored in image memories 60_1, 60_2, 60_3. An image comparing processor **62** combines corresponding resultant images, e.g. forms ratios of the images, to generate flow or perfusion enhanced images which are stored in a processed image memory **64**. An image processor **66** converts the contrast enhanced images into appropriate format for display on a monitor **68**, such as a CCD display, active matrix monitor, video monitor, or the like.

With reference now to FIGURE 2 along with continued reference to FIGURE 1, in a typical SS-EPI timing sequence for recording a T_2 weighted image according to the prior art, a 90° RF excitation pulse **100** is applied in conjunction with a slice selective gradient pulse **102** and an opposite polarity rephasing gradient pulse **104** which compensates for dephasing.

After the RF excitation is applied, there is typically a delay period τ_{dead} **106** whose duration is selected to allow partial T_2 relaxation. Thus, the duration of τ_{dead} **106** is preferably keyed to the T_2 behavior of the imaged tissue or contrast agent. Near the end of the delay period τ_{dead} **106** an initial phase encoding gradient pulse **108** and an initial frequency encoding gradient pulse **110** are applied. A 180° RF inversion pulse **112** is applied in conjunction with a second slice selection gradient pulse **114**.

The phase and frequency encoding directions are next cycled so as to traverse k-space and perform the SS-EPI encoding/readout operations. Only a few (about 40) phase and frequency encoding gradient pulses **120** and **122**, respectively, are shown in the exemplary qualitative timing sequence of FIGURE 2. However, it is to be appreciated that in typical SS-EPI imaging a larger number

of pulses are used, for example, 128 or 256 pulses, in order to obtain higher resolution. The SS-EPI encoding and readout timing sequence includes an oscillatory frequency encoding or read gradient **122** which encodes in the readout direction, preferably corresponding to the k_x coordinate in k-space, in a rastering back-and-forth manner. Synchronized with this oscillatory encoding gradient **122** are periodic phase encoding pulses **120** which step through the k_y coordinate in k-space. The phase and frequency encoding combine to spatially encode in-plane positions in the slice. A readout period τ_{read} **124** is located essentially centered about time-to-echo T_E **128**.

It is to be appreciated that the timing sequence of FIGURE 2, as well as those of FIGURES 3-6 which follow, are exemplary and qualitative sequences. It will be appreciated by those of ordinary skill in the art that additional pre-pulses (not shown) are optionally added to the sequence shown in FIGURE 2, or to any of the forthcoming timing sequences of FIGURES 3-6, to introduce additional contrast such as enhanced T_1 weighting, fat suppression, MTC, et cetera. Similarly, a partial parallel imaging (PPI) technique such as SMASH, SENSE, or PILS, is optionally incorporated into any or all of the echo planar readout waveforms discussed herein.

For a T_2 weighted image a dead time τ_{dead} **106** of order 40 msec is typical, corresponding to a time-to-echo T_E **128** of about 80 msec. The SS-EPI readout scan time τ_{read} **124** is typically of similar magnitude, e.g. less than 300 msec. Thus, the dead time τ_{dead} **106** is a significant fraction of the total scan time, and the sampling duty cycle for T_2 weighted SS-EPI imaging can be as low as 50%.

With reference now to FIGURE 3, a timing sequence is shown in accordance with one embodiment of the invention. A T_2 weighted image and a second image, which can be ρ weighted or T_2^* weighted, are recorded during a single repetition period T_R. The timing sequence includes

a 90° RF excitation pulse **200** in conjunction with a slice-select gradient pulse **202** and an opposite polarity rephasing gradient pulse **204** which compensates for dephasing, along with a 180° RF inversion pulse **212** and corresponding slice select gradient **214**. The time between the excitation and inversion pulses **200**, **212** is filled by a second image recording period. The initial phase encoding gradient pulse **208** and the initial frequency encoding gradient pulse **210** are applied after the RF excitation pulse **200**. The first readout begins thereafter. Data lines for a first image are acquired prior to the 180° RF inversion pulse **212**, i.e. during the corresponding dead time τ_{dead} **106** of FIGURE 2. The phase encoding gradients are applied such that data acquired around a time $T_{E,1}$ **216** is at the center of k-space. Depending upon the timing of the first time-to-echo $T_{E,1}$ **216**, this first image is relatively ρ weighted or T_2^* weighted. The 180° RF excitation pulse **212** is then applied in conjunction with the slice selection gradient pulse **214** and the second, T_2 weighted image is acquired about a time-to-echo $T_{E,2}$ **218** which is preferably centered on the spin-echo induced by the inversion pulse **212**.

In recording the two images, it will be appreciated that the phase stepping of the first image uses phase-select gradient pulses **220** with the same polarity as the phase stepping pulses of the second image **222**. However, because the 180° RF pulse **212** reverses the accumulated phase of the resonance spins, the k-space is effectively re-traversed in the same k_y direction for the two images.

Once again, it is to be appreciated that the timing sequence of FIGURE 3, as well as those of FIGURES 2 and FIGURES 4-6, are exemplary and qualitative sequences. Pre-pulses can be optionally included to introduce additional contrast such as enhanced T_1 weighting, fat suppression, MTC, et cetera. Partial parallel imaging (PPI) techniques can also optionally be

incorporated into the readout of the first image, the second image, or both images, using known techniques such as SMASH, SENSE, PILS, et cetera.

The timing sequence of FIGURE 3 provides additional image data over the prior art contrast enhanced imaging methods for the same imaging time. The additional data can be used, for example, to compensate for non-T_2 components of the T_2 weighted image. Such correction can be done by comparing the two images in a T_2-T_2^* analysis or a T_2-ρ analysis. In this manner, a more purely T_2 weighted image can be obtained, thereby facilitating accurate quantitative analysis of the perfusion of the gadolinium chelate **44** (FIGURE 1). It is to be appreciated that the method illustrated in FIGURES 1 and 3 overcome the prior art time constraint limitations on the collection of such additional data during the dynamic perfusion process by advantageously using the dead time in the prior art T_2 weighted SS-EPI timing sequence (FIGURE 2) to record the additional data.

With reference now to FIGURE 4 along with continuing reference to FIGURE 1, a timing sequence is described that generates data lines for ρ, T_2^* and T_2 images during a single repetition period T_R. The timing sequence of FIGURE 4 is similar to that of FIGURE 3, including an initial RF excitation pulse **300** with a corresponding slice-select gradient pulses **302**, **304**, similarly positioned initial phase encoding gradient pulse **308** and the initial frequency encoding gradient pulse **310**. The time is preferably lengthened between application of the excitation pulse **300** and a 180° RF inversion pulse **312** with a corresponding slice selection gradient pulse **314**. Data lines centered at $T_{E,1}$ **316** are sorted **50** into the ρ weighted k-space memory 52_1. Data lines centered at $T_{E,2}$ **318** are sorted **50** into the T_2^* weighted k-space memory 52_2. Data lines centered at $T_{E,3}$ **320** are sorted **50** into the T_2 weighted k-space memory 52_3.

In the timing sequence of FIGURE 4, it will be observed that the phase stepping of the ρ weighted image uses phase-select gradient pulses **322** of opposite polarity and equal magnitude and number compared with phase stepping gradient pulses **324** of the T_2^* weighted image. Thus, the same central portion of k-space is traversed in opposite k_y directions for the two images. Phase encode gradients **326** step the data lines back through the central region of k-space with the data line at time $T_{E,3}$ at the center of k-space. Data lines sampled after the 180° pulse remote from $T_{E,3}$ and phase encoded near an edge of k-space are optionally also sorted into the T_2^* or ρ weighted k-space. In this manner, the data lines away from the center of k-space are shared by two or by all three images.

Once again, it is to be appreciated that the qualitative timing sequence of FIGURE 4 can optionally include additional elements known to the art that produce additional contrast such as enhanced T_1 weighting, fat suppression, MTC, et cetera. Incorporation of partial parallel imaging (PPI) techniques into one or more of the image readouts can advantageously reduce the readout time and subsequently the overall imaging time.

With reference now to FIGURE 5, in another embodiment, two T_2 weighted images are recorded during a single repetition period T_R. This sequence is applicable to single shot T_2 mapping. An RF excitation pulse **400** is applied with a corresponding slice select pulse **402, 404**. An inversion pulse **412** and a corresponding slice selection gradient pulse **414** induce a first spin-echo. Data lines for a first T_2 weighted image corresponding to a time-to-echo $T_{E,1}$ **420** are sampled. Subsequently, a second 180° inversion pulse **416** with a corresponding slice selection gradient pulse **418** is applied to induce a second spin-echo. Data lines for a second T_2 weighted image corresponding to a time-to-echo $T_{E,2}$ **422** are then sampled during read gradient pulses of a second EPI segment. Of

course, additional cycles each including a 180° inversion pulse and oscillating read gradient pulses can be appended to the sequence, constrained by the length of the time constant T_2.

The timing sequences shown in FIGURES 3-5 preferably generate a full set of k-space data in a single T_R, i.e. single-shot EPI (SS-EPI). The speed of the SS-EPI technique permits, for example, real-time perfusion studies with corrected T_2 weighted image data. Of course, data can also be collected over several T_R.

With reference now to FIGURE 6, in yet another embodiment, two T_2^* weighted images are generated in a single acquisition period. Such a sequence is applicable, for example, to single shot T_2^* mapping, or to fat/water separation imaging.

An RF excitation pulse **500** is applied with a corresponding slice select gradient **502, 504** to induce resonance in a selected slice. An initial phase gradient pulse **505** preferably positions the magnetization out of the center of k-space in the phase encode direction.

An oscillating read gradient **506** induces a string of field echoes, each of which is sampled to generate one data line of k-space. Phase encode gradients **508** applied before each data line is sampled step the data lines through k-space. These data lines, which are temporally centered at $T_{E,1}$ **510** are reconstructed into an image with predominant ρ density weighting. The polarity of the phase encode gradient pulses **508** is reversed **520** and the data lines are stepped back through k-space. The data collected prior to the phase encode gradient rewind is more purely proton density weighted and those after the phase encode gradient reversal are more heavily T_2^* weighted. Images reconstructed from the data sets centered on $T_{E,1}$ **510** and $T_{E,2}$ **512** are combined by the processor **64** to emphasize T_2^* weighting effects and remove the effects of proton density weighting.

With continuing reference to FIGURE 6 and with further reference now to FIGURE 7, some data lines are shared by the two images. FIGURE 7 shows the sampling of k-space **600** during the EPI multiple echo data acquisition whose timing diagram is shown in FIGURE 6. The collection along the k_x direction rasters back-and-forth. Before each forward or backward pass a phase encode pulse **508** is applied to step the acquisition in the k_y direction. In a data acquisition portion **602** around $T_{E,1}$ that corresponds to an unshared k-space region **604**, this k_y stepping occurs in single-steps. This data is sorted into the ρ weighted k-space memory 52_1.

A second data acquisition portion **606** is phase encoded in a shared, peripheral k-space region **608**. Here, the k_y stepping occurs in double-steps through the application of higher strength phase encoding pulses 508_1. As the acquisition continues in the $-k_y$ direction, the odd k_y values (-5, -7, ... -15) are skipped and only the even k_y values (-6, -8, ... -16) are acquired. At a $-k_y$ edge **610** of k-space **600** the phase select pulse polarity reverses **520** and k-space is traversed in the $+k_y$ direction. The data obtained from the second portion **606** are stored in both the ρ weighted k-space memory 52_1 and the T_2^* weighted k-space memory 52_2, i.e. the data is shared.

A third data portion **612** is acquired in the shared k-space region **608**, with the acquisition now proceeding in the $+k_y$ direction. The odd k_y values are now acquired in the $+k_y$ direction (-15, -13, ... -5). The third data portion **612** is again recorded in both the ρ weighted k-space memory 52_1 and the T_2^* weighted k-space memory 52_2.

Echoes in a fourth data portion **614** around $T_{E,2}$ are phase encoded in single steps near the center of k-space, i.e. in the k-space region **604**. The data lines from the sequence portion **614** are sorted into the centered, unshared region of the T_2^* k-space memory 52_2.

An unsampled region 616 of k-space is filled by data lines that are conjugately symmetric to data lines in the shared k-space region 608.

Data sharing can advantageously be combined with optional pre-pulses and other sequence elements known to the art that produce additional contrast such as enhanced T_1 weighting, fat suppression, MTC, et cetera.

Incorporation of partial parallel imaging (PPI) techniques along with data sharing for one or more of the image readouts can further reduce imaging time.

In one preferred embodiment, the method improves upon the prior art by providing for acquisition of data in addition to the usual SS-EPI T_2 weighted scan. The additional data is used to generate more purely T_2 weighted data. A second scan (T_2^* or ρ weighted) is acquired and is quantitatively compared with the T_2 weighted scan to produce a third image which has a purer T_2 weighted contrast. For example, a T_2-T_2^* or a T_2-ρ analysis is preferably performed. The multiple echo SS-EPI is preferably used in conjunction with a contrast-enhanced imaging experiment, such as a dynamic perfusion experiment using a gadolinium chelate contrast agent that reduces the T_2 contrast. Of course, other contrast agents can also be used in accordance with the method. Based upon the improved T_2 contrast of the third image which is calculated from the first and second measured images, a quantitative determination of the perfusion of the contrast agent is obtained. This perfusion determination can be obtained by taking ratios or logarithms of the two images in the image comparison processor 62 to quantify the time course and arrival curves. Given a predetermined relationship between the contrast agent concentration and the reduction in T_2 contrast (e.g., an empirical relationship), the concentration distribution of the contrast agent is determined with respect to spatial and temporal coordinates. Thus, for example, the perfusion of the

gadolinium chelate are quantitatively determined based upon the third image.

In another embodiment, a temporal evolution of a first clinical parameter is obtained from a temporal series of images obtained using a first echo planar readout waveform. A temporal evolution of a second clinical parameter is obtained from a temporal series of images obtained using a second echo planar readout waveform, the first and second readout waveforms being included in a single multiple-contrast EPI imaging sequence. The temporal evolution of the first and second clinical parameters are combined, e.g. mathematically or by qualitative interpretation by medical personnel, to obtain additional diagnostic information that is unavailable from prior art methods. Alternatively, the images obtained from the first and second readout waveforms are combined to form a third image, and the temporal evolution of a parameter related to the third image is then extracted.

In yet another embodiment, a first scan of a multiple echo enhanced contrast SS-EPI experiment provides a high contrast image which is responsive to low concentrations of the contrast agent, but which is effectively overexposed in regions of high concentration such as the major blood vessels. A second image reconstructed from the sequence preferably is a medium contrast image which clearly shows the contrast agent in the major vessels but does not resolve the lower concentration perfusion areas as well as the first image. The two images together provide a medical doctor with more complete information than is provided by either image by itself. The contrast agent is again preferably a gadolinium chelate, although of course other contrast agents can also be used in accordance with the method.

In still yet another embodiment, a first image of a multiple-echo enhanced-contrast SS-EPI experiment is tuned to be highly sensitive to the contrast agent. In

the exemplary case of a gadolinium chelate contrast agent, this first image is preferably a T_2 weighted image. A second image of the sequence is preferably recorded under conditions which are not sensitive to the contrast agent. For the gadolinium chelate, a ρ weighted image is appropriate. In this way, the first image of the contrast agent is placed into context by the second image, which serves as a reference image. The images are both acquired during a single repetition period T_R, so that subsequent registry is unnecessary. Additionally, any movement of the patient during the dynamic perfusion experiment is automatically included in both the contrast enhanced images and the reference images.

The invention has been described with reference to the preferred embodiments. Obviously, modifications and alterations will occur to others upon reading and understanding the preceding detailed description. It is intended that the invention be construed as including all such modifications and alterations insofar as they come within the scope of the appended claims or the equivalents thereof.